AD-769 022

DYNAMIC CHARACTERISTICS OF THE HUMAN BODY

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1973

AD 769 022

Security Classification

DOCUMENT CONTROL DATA - R & D

(Security classification of title, body of abstract and indexing annotation must be entered when the overall report is classified)

1. ORIGINATING ACTIVITY (Corporate author) Aerospace Medical Research Laboratory Aerospace Medical Division, Air Force Systems Command Wright-Patterson, AFB, Ohio 45433	2a. REPORT SECURITY CLASSIFICATION Unclassified 2b. GROUP

3. REPORT TITLE

DYNAMIC CHARACTERISTICS OF THE HUMAN BODY

4. DESCRIPTIVE NOTES (Type of report and inclusive dates)

5. AUTHOR(S) (First name, middle initial, last name)

H.E. von GIERKE

6. REPORT DATE	7a. TOTAL NO. OF PAGES	7b. NO. OF REFS
1973	10	21

8a. CONTRACT OR GRANT NO. b. PROJECT NO. 7231	9a. ORIGINATOR'S REPORT NUMBER(S) AMRL-TR-72-53 9b. OTHER REPORT NO(S) (Any other numbers that may be assigned this report)

10. DISTRIBUTION STATEMENT

Approved for public release; distribution unlimited.

11. SUPPLEMENTARY NOTES	12. SPONSORING MILITARY ACTIVITY
Reprinted from PERSPECTIVES IN BIOMEDICAL ENGINEERING. R.M. Kenedi,ed.	Aerospace Medical Research Laboratory Aerospace Medical Division, AFSC, Wright-Patterson AFB, Ohio 45433

13. ABSTRACT

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DYNAMIC CHARACTERISTICS OF THE HUMAN BODY

H. E. VON GIERKE[1]

INTRODUCTION

Most studies of the mechanical properties of the human body have been conducted on isolated organs, body segments, limbs or tissue specimens. For these the static and dynamic functions under internal muscle loads or external forces, the geometry and structure are relatively well defined. Results of this type have reached a high degree of sophistication using the whole framework of modern engineering dynamics and stress analysis in the course of the studies. However, if we are interested in the behaviour of the whole body or larger segments of it and are not willing to restrict ourselves to specific force inputs or loads, the number of studies conducted and the information available become less and less. The reasons for this are at least threefold: (1) the overall system becomes extremely complex and its mechanical functions and abuses very manifold. Although engineering systems of high complexity; as for example aircraft, are analysed by dynamic models with respect to their dynamic responses and stress loads down to their individual subsystems and components, they have the advantage of being built of materials with known and understood material properties. (2) Testing of the overall system is almost impossible with simultaneous detailed measurements of the responses of subsystems and components. This is particularly true for the human body *in vivo* and to a large extent even for cadaver materials. (3) Of the many factors determining the dynamic characteristics of the human body (table 1) at least half of them are not constant but vary with time. Several of these parameters determining the body's response are interrelated; for example, the time course of the force input function and the body's geometry determine jointly how the mechanical energy is propagated through the tissue (figure 1). Many of the difficulties listed in table 1 could be overcome by a more genuine interdisciplinary approach and, somewhat connected with this, by posing the overall problem in broader terms leading to results of more general validity. There is no doubt that in the long run this broader attack would be the more rewarding and more economical one. It is hoped that the following brief, but nevertheless critical, review of our knowledge will help to outline some of these proposed more general attacks on the problem.

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THE PURPOSE OF INVESTIGATIONS IN HUMAN BODY DYNAMICS

Most investigations in human body dynamics (table 2) were conducted to explain the injuries, performance decrements or sensations caused by periodic or impulsive force environments, or to develop protective equipment against these environments. The body's response in the reversible linear and nonlinear loading range was of primary interest in most cases. However, the injury mechanisms in what are termed 'environmental exposures' in the table involve determining where various force loadings produce primary tissue damage and at what input levels to the body such damage occurs. Although knowledge of the basic tissue strength properties of isolated specimens is helpful in this task, they cannot—or at least not yet—replace experimental studies or accident data analysis on the composite structure or its analogues. It is important to note that always several of the methodologies reviewed in the following must be applied to shed light on body dynamics of interest for one of the purposes listed. When in table 2 the purpose of some investigations with periodic or impact forces is classified as 'methodology', it is meant that the excitation force function as such was of no direct interest but that the data gained with this test added information on body dynamics of interest in connection with other purposes or force excitations.

Dynamic characteristics of body	Depend on	Soft tissue properties
		Bone properties
		Soft tissue and bone geometry
		Body position/support/restraint
		Muscle tension
		Preload on tissue
		Force input time function/place

TABLE 1: Factors influencing dynamic characteristics of the human body

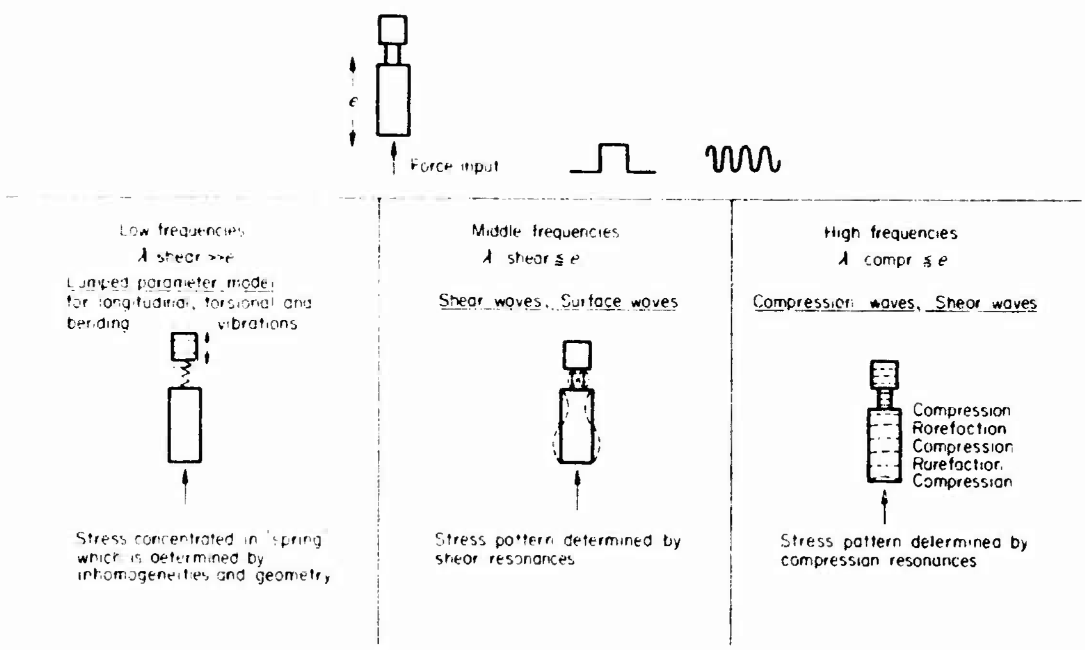

Figure 1. Different modes of propagation of mechanical energy in body tissue. Depending on the time function (frequency content) of the force input the energy is propagated in the form of transverse shear waves or longitudinal compression waves. Stress concentrations depend also on inhomogenities and geometry (λ = wavelength) (from von Gierke,1964).

Periodic excitation	Environmental exposure (vibration, noise)	Injury
		Physiological effects
		Performance effects
		Protection
	Physiology	Haemodynamics
		Respiration
		Locomotion
	Methodology	Tissue/component properties
		Relation to impulse excitation
		Nonlinear properties (static + oscillatory loading)
Impulse excitation	'Environmental' exposure (crash, aircraft escape, blast)	Injury mechanisms
		Protection
	Physiology	Pulse wave
	Methodology	Shorten oscillatory studies

TABLE 2: Purpose of investigations on dynamic characteristics of the human body

METHODOLOGIES

The various approaches to analysing and understanding the human body's dynamic response are listed in table 3. Most of the methodologies have been applied with impulsive as well as oscillatory forces and have been used either to solve specific applied problems or to contribute to the basic overall goal. It is only fair to state that none of the methods by itself can come close to a solution of the question asked: how does the human body respond to mechanical forces, in the elastic as well as the injury range? Only the combination of the results obtained by all approaches can bring us closer to this goal. And it is here, where perhaps the greatest challenges and problems are, where the last methodology listed, biodynamic modelling, takes on a special position and importance with respect to the others. Not that it could give us a realistic answer without all the other methods—on the contrary, it is entirely dependent on the experimental input from all the other methodologies. But it is the only one which contains in itself the hope of providing the proper framework to combine, correlate, understand and use the total body of information available in this field. In my opinion, this is the area in which the major progress has been achieved in the last decade and which should guide our endeavours in the future (von Gierke, 1971a). This does not mean that it is the area where the major

In vivo human body response	Kinematics
	Force velocity transfer functions
	Driving point impedances } { Whole body
	Transfer impedances } { Body segments
	Volume impedances
	Physiological responses
	Performance
	Accident analysis
Cadaver response	Kinematics
	Injury mechanisms and levels
	Component analysis
Animal response	Kinematics
	Force velocity transfer functions
	Impedances
	Physiological response
	Performance
	Component analysis
Biodynamic model response	Kinetic models
	Total body response models
	Subsystems models
	Tissue property models
	Animal models scaling laws
	Hardware models

TABLE 3: Methodologies for analysing dynamic whole body characteristics

work or resources are required, but that it must form the foundation, without which the other investigations remain fragments. For this reason the various types of models which resulted from the experimental investigations will be taken as the framework for the following discussion of the present status of the field.

BRIEF REVIEW OF THE BODY'S DYNAMIC RESPONSE

In most cases the effects of mechanical energy on man are the direct result of relative displacement of body parts or segments, organ deformation or local tissue strain. This is certainly true for vibratory as well as impulsive force inputs as far as acute as well as chronic tissue damage and injury are concerned. It is also the case for most known interference effects of vibration with human performance such as manual control (Allen *et al.*, 1972), speech (Nixon *et al.*, 1963), or visual acuity (O'Briant *et al.*, 1970) and also must be assumed to account in a more diffuse and complicated way for such integrated effects as fatigue, discomfort or those effects on performance we are not able to relate directly to head, arm or hand motion (von Gierke, 1971*b*). Correlations of the latter effects with causes other than the motions of single body parts have not yet been tried, with the exception of the probably oversimplifying assumption to relate them to the total mechanical power dissipated in the whole body. The measurement and description of tissue strain suffers from the inaccessibility of living tissue to these types of measurements, the lack of appropriate landmarks and the difficulty of quantitating all those factors listed above as influencing the body's dynamic response. As a consequence the body's response is best described and understood in terms of biodynamic models which are compatible with available measurements and, at the same time, elucidate which information is still missing (von Gierke, 1971*c*; Sandover, 1971).

Kinetic body response

The gross motions of individual body segments or parts have been studied in internal biomechanics; i.e. as a result of internal energy capabilities (motion, force and torque, gait, etc., as a result of muscular strength, work, power) and in external biomechanics, i.e. under the influence of environmental forces (translation or rotation of the body under blast forces, flailing of head or limbs in the airstream following emergency escape from aircraft and primarily kinetics of aircraft or automobile occupants under crash conditions). Based on kinematic observations kinetic models of the stick-and-joint type or with the proper anthropometric parameters of the individual body segments are constructed. An example of such a model is shown in figure 2. Frictional constraints and limitations of joint articulations are based on empirical fits; however, no attempts are made to relate joint forces with any actual forces acting inside the body or causing injury to body structures. In these models, the elastic behaviour of joints is also neglected, so that oscillatory body responses cannot be realistically calculated. In spite of these limitations, which make these descriptions appear primitive from a biological point of view, these models are extremely helpful for the prediction of body motions as a result of forces such as occurring in automobile

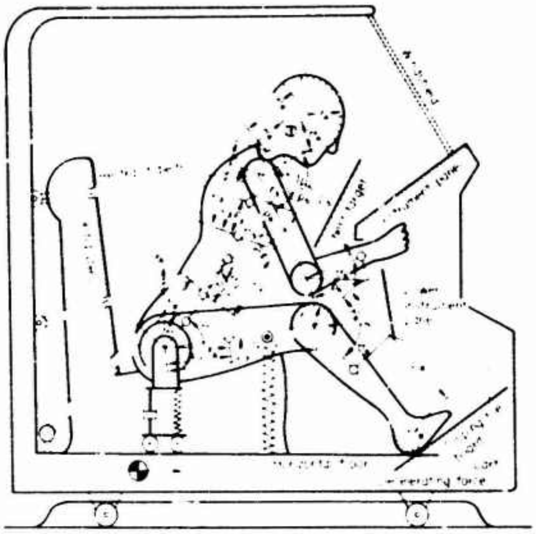

Figure 2. Kinetic model of the human body and restraint system on a test cart (11 degrees of freedom) (from McHenry, 1971).

crashes and to calculate time course, magnitude and location of impact forces to individual body parts such as head, chest or knee. In this context they are of great value in the evaluation of restraints and other protective systems, and in parametric studies of inflatable safety restraints (air bags), etc. In internal biomechanics, kinetic models of this type are used to study walking mechanisms or to calculate positions of body members, of force, lift and torque capabilities incorporating data on isometric muscle strength or other energy capability assumptions. A combination of external and internal kinetic models has been attempted only by including to a partial degree the environmental effects of g-fields into internal models of motion capabilities (Huston *et al.*, 1971). Future developments in this field are foreseen to include time-varying muscle forces and some sort of active closed loop motion control into models of the type shown in figure 2.

Oscillatory body response

Under vibratory force inputs the body exhibits resonances in various frequency ranges which lead to specific physiological manifestations, performance decrements, or injuries. To approximate these complex body deformations and wave propagations, ratios of vibration amplitudes at various body positions have been measured, as for example, the ratio of hip to table, shoulder to table, or head to table amplitude of a sitting subject vibrated on a shake table (figure 3). The second type of measurement relates to volume changes in abdominal circumference, chest circumference and air compressed and forced out of the lungs (figure 4). The third type of measurement frequently reported is the input impedance of the human body at the area of force transmission (the buttocks for the sitting

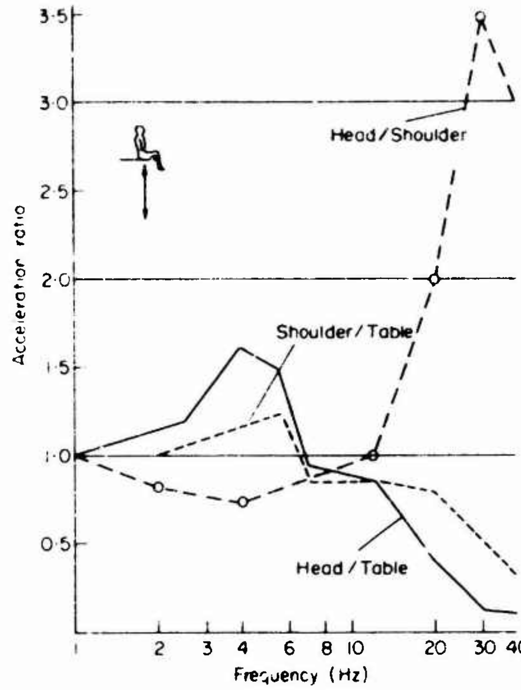

Figure 3. Vertical (Z-axis) vibration transmitted to the sitting human subject (from Dieckmann, 1957).

subject) (figure 5). The combined results of these measurements can be understood and described in terms of total body response models of the lumped parameter type shown in figure 6. Representative model parameters used with this model are given in table 4. These parameters must agree not only with the results of the three types of measurements mentioned above but also with the data resulting from static anthropometry and the data available from detailed subsystems analysis. This is particularly true, for example, in the case of the spinal system with the abundant theoretical and experimental work on the isolated spine, or in the case of the lung-thorax system with the available body of physiological information on respiratory

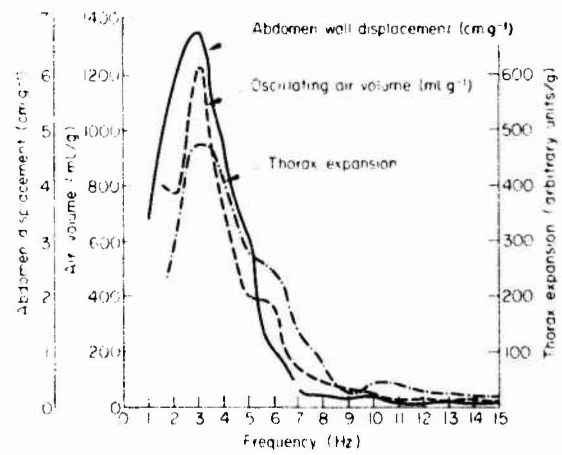

Figure 4. Abdominal wall displacement, thorax expansion and air volume oscillating through the mouth per g longitudinal vibratory acceleration (subject in supine position) (from Coermann *et al.*, 1960).

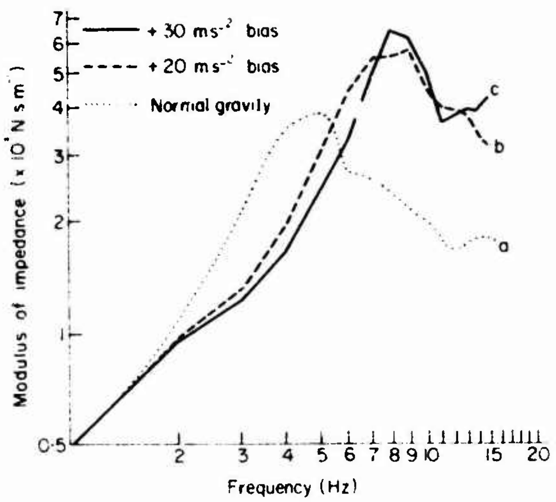

Figure 5. Mechanical driving point impedance of the sitting human subject subjected to *Z*-axis vibration at various static sustained acceleration levels on a centrifuge (From Vogt *et al.*, 1968); (a) under normal gravity, (b) under +2G$_Z$, (c) under +3 G$_Z$ acceleration.

V_o (cm^3)	4×10^3
A_r (cm^2)	2
M_r (gm)	1×10^{-1}
D_r (dyn s/cm)	$1{\cdot}6 \times 10$
K_r (dyn/cm)	0
A_w (cm^2)	2×10^2
M_w (gm)	1×10^3
D_w (dyn s/cm)	6×10^5
K_w (dyn/cm)	1×10^8
A_a (cm^2)	2×10^2
M_a (gm)	4×10^3
D_a (dyn s/cm)	1×10^4
K_a (dyn/cm)	8×10^6
M_t (gm)	4×10^4
D_t (dyn s/cm)	4×10^6
K_t (dyn/cm)	1×10^9
M_p (gm)	8×10^3
D_b (dyn s/cm)	$6{\cdot}5 \times 10^5$
K_b (dyn/cm)	6×10^7

TABLE 4: Typical model parameters for system shown in figure 6.

mechanics. Therefore the fitting of the parameters to all available data is an extremely difficult job. It must be kept in mind that the model will never tell us more than the type of measurements on which it is based. It can explain these measurements, it can make many additional measurements of the same type unnecessary, and can be all around an extremely useful tool; however the assumptions on which it is based should never be left out of sight. For example, it is quite obvious from an analysis of the system in figure 6 that primarily the buttocks system is reflected in driving point impedance measurements at the buttocks. Although such measurements are very useful to obtain the total mechanical energy transmitted from the seat to the man, they cannot tell us in such a series system where the energy goes and what responses it elicits there (Payne *et al.*, 1971). If the impedance measured at the buttocks can be represented fairly well by a one or two degree of freedom system, it is erroneous to assume—as it has unfortunately been done—that there are not important separate additional degrees of freedom such as the spinal subsystem or the lung-thorax system, which are not reflected in the buttocks input impedance. Of the measurements possible on the outside (at the surface) of such a system, it appears as if one type of measurement has been neglected or hardly been tried to our knowledge: the input-output impedance or transfer impedance. Treating the system—and also its various subsystems—as four pole, additional information could be obtained by either loading (for the buttocks input case) the torso mass or the chest wall output with known impedances (mass loading), or by trying to reverse input and output; for example, by driving the torso mass at the shoulders and by measuring the input impedance there and the output impedance at the buttocks. It is somewhat unfortunate that the impedance measurements on the sitting human subjects done for the first time more then 30 years ago have been extensively repeated and misinterpreted, but that not too many new and imaginative methods have been added.

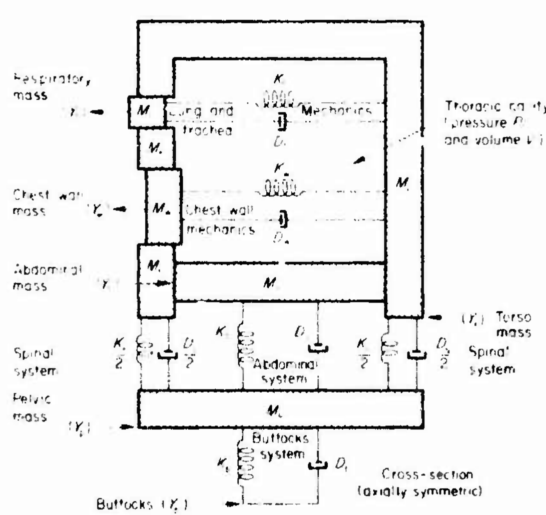

Figure 6. Example of a total body response model for the frequency range 1 to 100 Hz. The model is used to calculate body deformation (spinal compression, pressure in the lungs, etc.) as a function of external longitudinal dynamic forces (vibration or G_Z impact) and pressure loads (blast, acoustic pressure, decompression loads) (from Kaleps *et al.*, 1971).

Human impedance measurements as well as force-deflection measurements on the buttocks and on isolated spinal segments show a marked nonlinearity. With the exception of the driving point impedance functions, for which measurements on a centrifuge at various static preloads gave valuable information (Vogt *et al.*, 1968) (figure 5), these nonlinearities are largely uninvestigated and deserve further study and definition. A detailed analysis of available data led

Payne *et al.* (1971) to propose the nonlinear spring constants shown in figure 7 for buttocks and spinal spring, which are in agreement with all available measurements on nonlinearities. The linearised four mass system of figure 7 results in the amplitude ratio curves in figure 8, which are similar to the measured ones in figure 3. The lung-thorax system is

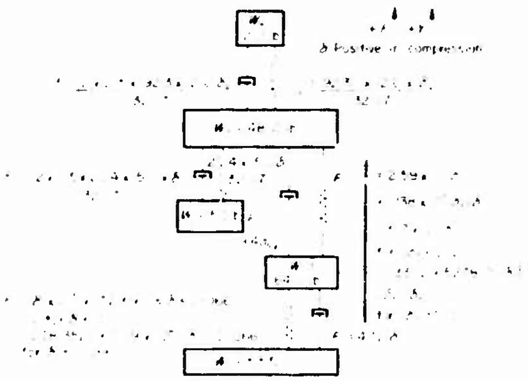

Figure 7. Four degree of freedom lumped parameter model with nonlinear spinal and buttocks characteristics. The model describes longitudinal impact and vibration response (from Payne *et al.*, 1971).

omitted from this model which is being used primarily to study spinal response. How much, however, the absolute magnitude of the transmission factor and of its frequency response depend on the body posture and how much such changes in the frequency response affect the model representation needed to approximate these frequency responses are illustrated in figure 9 for the example of the table to head transmission (Potemkin *et al.*, 1971). If pressure in the lungs and lung damage are of

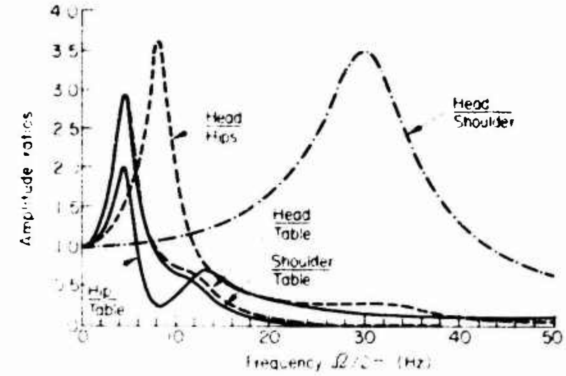

Figure 8. Amplitude transmission ratio for the model of figure 7 (linearised). Compare to measured transmission ratios of figure 3 (from Payne *et al.*, 1971).

primary interest, as in the case of blast or infrasound effects, the pressure-volume relationship for the air in the lungs cannot be linearized as is usually done for approximating the whole body oscillatory case (Kaleps *et al.*, 1971; Bowen *et al.*, 1968).

In addition to the body deformations under oscillating forces so far described, there have been effects observed on the cardiovascular system caused by sinusoidal accelerations (Edwards *et al.*, 1972). In anaesthetised dogs these changes in peak aortic flow were in the 3 to 9 Hz range, and during 3 g vibration at 4 Hz amounted to a maximum aortic peak flow rate of more than twice the control value and a minimum aortic peak flow rate of 10% of the control value. The amplification or reduction of peak flow rate depended on the phase relation between vibration and cardiac cycle. The pulse pressure under these conditions could reach

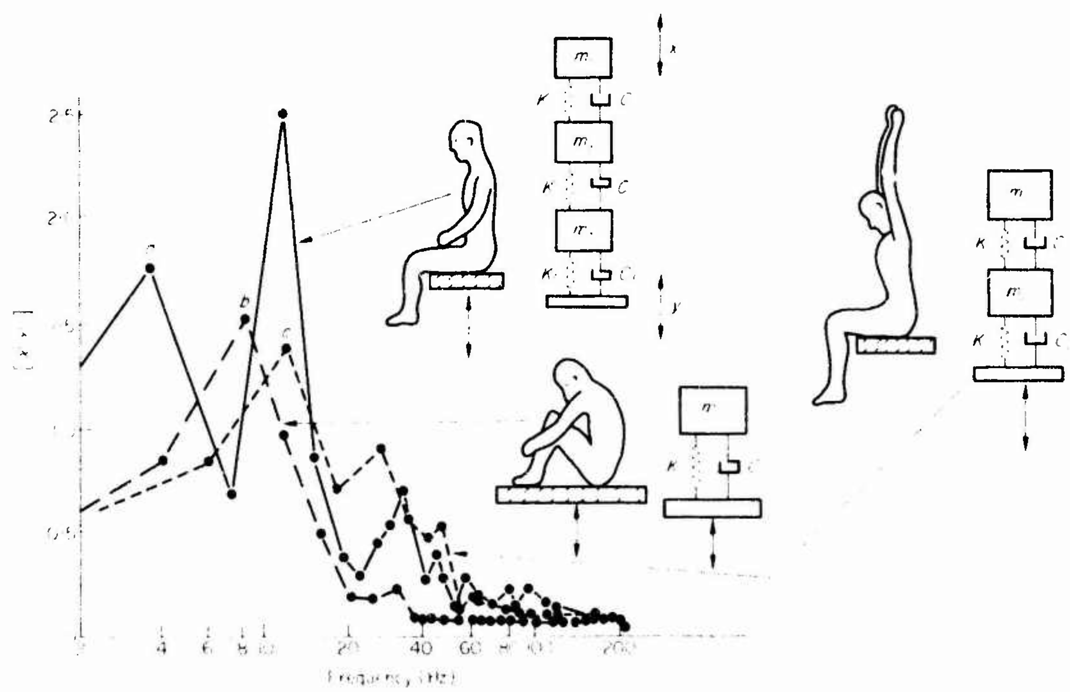

Figure 9. The square of the table to head transmission ratio as a function of frequency and body posture. The networks needed to approximate the functions are also indicated (from Potemkin *et al.*, 1971).

more than five times the control value. Without considering further the physiological implications of these phenomena, which have not yet been studied in humans, it is of biomechanical interest to note that these findings have been explained by modelling the hydraulic aspects of the circulatory system alone; i.e., these effects are apparently not caused by deformation of the vessels because of the mechanical resonance of the abdomen and the chest or by changes in physiological feedback mechanisms. Consequently, it will be necessary to incorporate the model of the cardiovascular system under sustained acceleration and vibratory stress, into the whole body response model discussed above. Such a refined model would have very promising applications to environmental physiology (circulation under hypo- and hypergravic conditions as well as under vibration and impact stress) as well as to problems in clinical medicine (circulatory assist by alternating, heart synchronised forces or pressures).

Impact response

As long as our considerations are restricted to the direct physical effects of the mechanical force environment, there is a clear, well known mathematical relation of the vibration response of the human body to the impact response. This relationship is schematically indicated in figure 10. The left-hand side shows the overall human vibration 'tolerance' curve as the composite of curves of equal tissue strain for various subsystems; i.e. tissue areas. In principle each curve of equal tissue strain under vibration stress on the left-hand side, corresponds to a curve of equal tissue strain under impact stress on the right-hand side of the figure. The transmission ratios observed in human impact acceleration experiments, as well as the data obtained from accident analyses, support the general trend of these theoretical impact tolerance curves. However, quantitatively the relationship is considerably more difficult due to the fact that the main interest in curves of equal tissue strain for the impact case is at stresses beyond the linear range of the system, namely, in the tissue loading range close to or at the point of irreversible damage (Payne *et al.*, 1971). For this reason the nonlinear lumped parameter model of figure 7 reflects much better experimental data in the impact range of minor probability of spinal or abdominal injury than the linear model of figure 6. Nevertheless, the linear model of figure 6 correctly explains observations, in accident investigations and animal experiments, that for short duration acceleration pulses spinal compression fracture is the most sensitive injury mechanism whereas for longer duration pulses of the same magnitude abdominal injury is more likely to occur (figure 11).

The preceding discussion focused on environmental loads in the direction of the Z-axis of a sitting subject only. The response dynamics are naturally different for the other input directions and different model systems must be designed to fit the impedance, transmission ratios, accident data and subsystem studies. Although a considerable body of information for the other axes is available, the information and its condensation into quantitised models are not yet so complete as for the Z-axis. For X-axis excitation (front to back) impedance

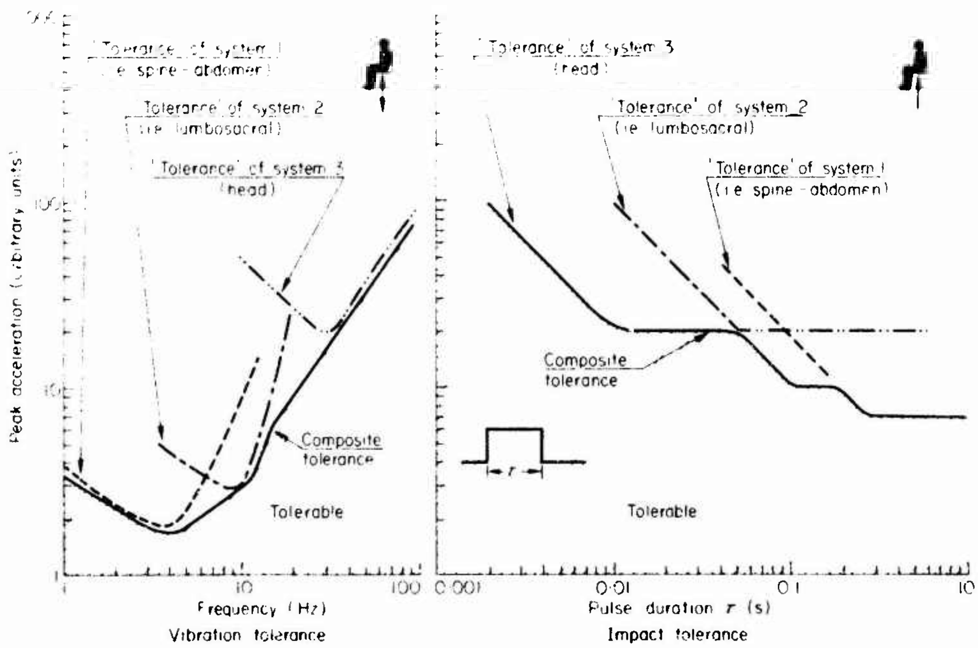

Figure 10. Relationship between vibration tolerance and impact tolerance of a sitting human subject (schematic). The dotted lines indicate lines of equal tissue strain in localised tissue areas; i.e., tolerance lines for the individual subsystems. The composite tolerance curve is the envelope to the individual subsystem curves and represents the overall tolerance curve for the total organism (from von Gierke, 1964).

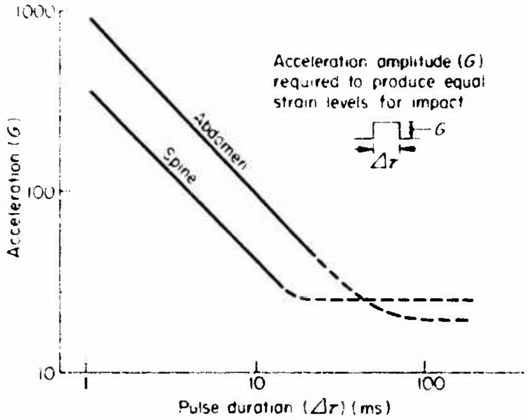

Figure 11. Curves of equal mechanical strain (equal probability of injury) for abdominal displacement and spinal compression in the model shown in figure 6. Note that the cross-over of these curves depends upon the shape of the force input (from Kaleps *et al.*, 1971).

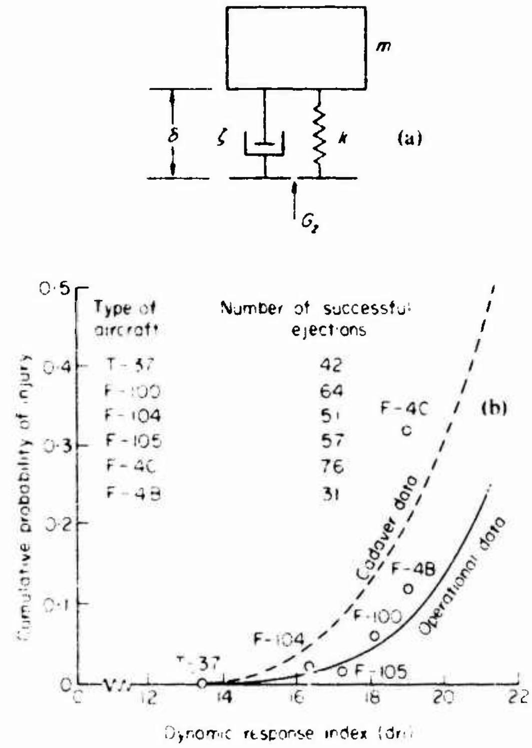

Figure 12. (a) Spinal injury model (from Brinkley, 1968). m is mass (lb s^2/in); δ is deflection (in); ζ is damping ratio; k is stiffness (lb/in); G_z is acceleration input (in/s^2); DRI $= \omega_n^2\delta_{max}/g$, where DRI stands for dynamic response index; $\omega_n = (k/m)^{1/2}$ (rad/s); and $g = 386$ in/s^2. (b) Probability of spinal injury predicted from cadaver data compared to operational experiences with various U.S. Air Force ejection systems (from Brinkley, 1968).

measurements even with static preload are available; however, the response dynamics leading to well defined accident mechanisms are inseparable from the body support (back and headrest) and restraint to permit deducing a generalised model.

APPLICATION OF DATA

The main advantage of presenting human body response data in terms of the biodynamic models discussed in not only the general understanding of the dynamic events elucidated by the model but primarily the application of the model to the solution of practical human tolerance and bioengineering problems. The interaction of man with his environment, for example, with the dynamics of his seat, seat cushions and restraints, is most satisfactorily analysed by means of such body response models. For the case of spinal compression injury as a result of longitudinal (Z-axis) loads as they occur during catapult emergency ejections from military aircraft, the statistical probability of injury is well enough known to correlate it with the strain in the spinal spring of the models in figure 6 or 7. If one limits the considerations to spinal injury only as the most sensitive injury mechanism in the short duration impact range (see figure 11), one can reduce the model to the oversimplified but practically very useful spinal injury model of figure 12. To specify for the design engineer permissible loads on the man in terms of permissible strain on the spinal spring (which in turn results in a predictable probability of injury) certainly makes much more sense scientifically, and in terms of operational data, than any definition of tolerance limits previously attempted. Needless to say, that for finer definition of the type and location of spinal injury a more refined model of the spinal subsystem than the one introduced here is required and desirable. But in general this way of presenting and using experimental data in an overall probability of injury model—which, if desired, can be backed up by a more refined subsystem injury model—should be looked at as a desirable goal for analysing and treating other injury mechanisms (Payne, 1971).

As an example of how the more complete model of figure 7 has been used to analyse the dynamics of the whole man/ejection seat system, the 'plane of symmetry' model (Band, 1971) derived from the model in figure 7 is presented in figure 13. It allows for calculating fore and aft movement of the man in the seat, includes restraining forces and allows for translational and rotational dynamics in the plane of symmetry (Z–X plane). An example of calculated forces transmitted at the buttocks, spine and neck during a catapult maneuver is illustrated in figure 14 and agrees well with the forces derived from photometric observation of human subjects during ejection loads.

The whole body response models are foreseen to be of similar usefulness in setting vibration exposure limits and analysing vibration isolation requirements once more information on the long term effects and possible chronic injury mechanisms is available and statistically documented. For these purposes the standardisation of 'nominal' human

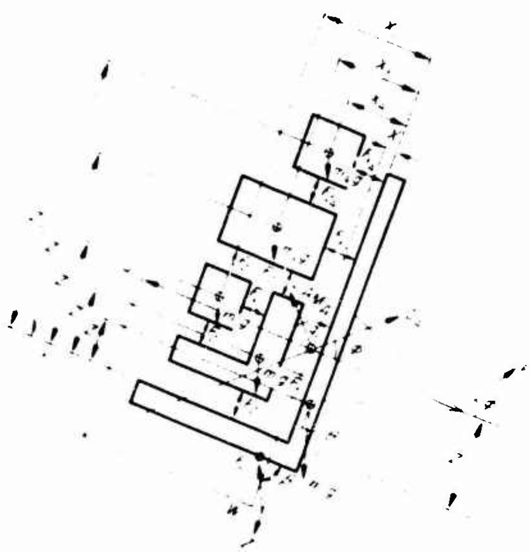

Figure 13. Eleven degree of freedom 'Plane of Symmetry' model of a rocket-powered, free-flight man seat system. For parameters indicated see reference below. F_R rocket thrust, F_D drag of stabilising drogue. D_A aerodynamic drag of man seat (from Band, 1971).

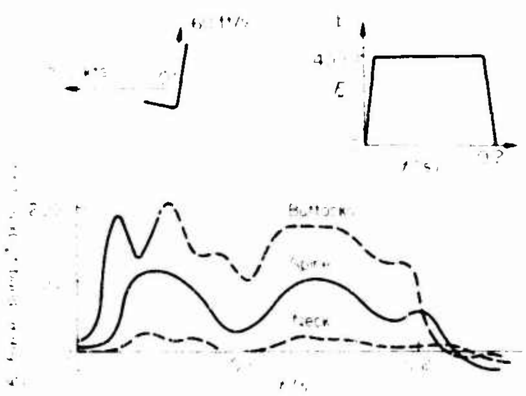

Figure 14. Forces as a function of time during catapult ejection of the model of figure 13. Initial conditions for seat velocities and rocket thrust F_R as indicated. Inclination of rocket thrust to X axis 60 (for details see Band, 1971).

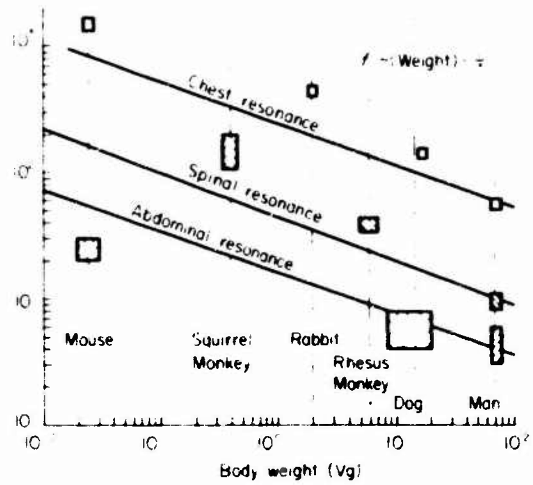

Figure 15. Approximate resonance frequencies of total body response models of the type shown in figure 6 as a function of body size (weight) (from von Gierke, 1971c). The shaded areas indicate ranges for measured data.

impedance values and transmission ratios is under consideration by various groups. These values are then foreseen to be used in the design of a new generation of anthropomorphic (or better, anthropodynamic) dummies, which incorporate the dynamic response characteristics of man for hardware, vibration and impact tests.

For the planning, interpretation and application of biodynamic data obtained on animal models, it is important to realise that to a first approximation the resonance frequencies increase as the linear body dimensions decrease (von Gierke, 1964, 1972). This fact not only shifts the frequency scale (figure 15) for comparable vibratory responses and the duration scale for comparable impact responses, but results, for the assumption of basically equal tissue strength, in the general conclusion that the smaller the animal the higher the vibration and impact loads leading to similar manifestations. Although experimental data clearly support these theoretical predictions, more detailed work on this subject is desired.

OUTLOOK

This short review could do little more than call attention to the broad scope of its title and sketch some of the typical and important developments in this field. It is hoped that it called attention to the following facts and predictions:

(a) future progress toward a unified, quantitative picture of man's mechanical response to dynamic force environments must come from a refined mathematical description of the biomechanical and biological processes. Experimental data collection should be based on the theoretical model framework available so that no important parameters required for their quantitative integration are overlooked or neglected. Representative statistical samples should be aimed at in describing physical response phenomena as well as physiological or injury mechanisms.

(b) It would be desirable to derive, on the basis of experimental data, a generally valid breakdown of the total body system into subsystems and sub-subsystems. The range of useful and valid application of each subsystem and the coupling between subsystems should be identified together with the biological phenomena hopefully to be described by the particular sub-model approach. For example if a kinetic model analysis of the type shown in figure 2 results in specified force impacts to the skull and the chest, application of a chest sub-model and head-neck sub-model would identify the dynamic responses and biological

effects of these specific impacts. After analysis of the head-neck dynamics, it might be necessary to go on to a further, still more refined head injury model, which allows differentiation of the probability of occurrence of the various types of head injury mechanisms. It appears as if a much more systematic approach to the analysis of the system structure must be taken than has been the case in the past. Conceptionally this breakdown should be continuous starting with the models describing the dynamics of the macroscopic body structure, characterised by anthropometric data, down to subsystem models describing basic elastic and strength properties of structural components and finally of the material itself.

(c) 'Active' responses of the body to the mechanical force environments should be incorporated into the passive response models discussed above.

In summary, the preceding discussion of the body's dynamic response to vibration and impact must remain an unsatisfactory accumulation of data until a more coherent effort in data collection, as well as theoretical foundation, allows more direct and uninterrupted correlation between the two. It appears as if the solution does not necessarily rest in an expansion of this research field but in a maturation process and a logical, truly interdisciplinary, step-by-step research programme.

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